

Finite-Element-Based Transfer Equations: Post-Mortem Human Subjects versus Hybrid III Test Dummy in Blunt Impact

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ABSTRACT

In the present study, transfer equations relating the responses of post-mortem human subjects (PMHS) to the mid-sized male Hybrid III test dummy (HIII50) under matched, or nearly-identical, loading conditions were developed via math modeling. Specifically, validated finite element (FE) models of the Ford Human Body Model (FHBM) and the HIII50 were used to generate sets of matched cases (i.e., 256 frontal impact cases involving different impact speeds, severities, and PMHS age). Regression analyses were subsequently performed on the resulting age-dependent FHBM- and HIII50-based responses. This approach was conducted for five different body regions: head, neck, chest, femur, and tibia. All of the resulting regression equations, correlation coefficients, and response ratios (PHMS relative to HIII50) were consistent with the limited available test-based results.

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INTRODUCTION

Risk curves are typically derived from statistical analyses of injury outcomes from tests involving post mortem human subjects (PMHS) and corresponding physical input variables such as force, compression, etc. Because test dummies such as the Hybrid III mid-sized male test dummy (HIII50) are designed to have acceptable biofidelity, their responses are often directly compared to PMHS-based risk curves. However, "transfer equations", which relate cadaveric responses to those of the HIII50 subject in matched, or nearly-identical, loading conditions, may potentially improve the fidelity and applicability of the risk curves.

Test-based transfer equations are often difficult to generate due to the limited availability of PMHS for matched testing. For example, Laituri et al. (2005) derived a PMHS-based thoracic risk curve in terms of age and normalized chest deflection. A transfer equation was needed to convert the PMHS-based risk curve into a HIII50-deflection-based risk curve. However, a literature review yielded only 20 approximately-matched tests, and many of those tests involved mixed loading types. Moreover, due to the small sample size, age effects could not be established from the matched tests. Laituri et al. subsequently applied the resulting, albeit limited, test-based transfer equation, but noted that supplemental math modeling might be beneficial.

In the present study, model-based transfer equations were developed. First, finite element (FE) models of the age-dependent Ford Human Body Model (FHBM) and the HIII50 were validated against physical tests. Next, the validated models were used to create a set of responses from matched cases involving different impact speeds, severities, and FHBM ages. The resulting FHBM and HIII50 responses were then subjected to regression analyses to generate transfer equations. This approach was applied in separate analyses of the head (n = 28 cases), neck (n = 6 cases), chest (n = 6 cases), femur (n = 18 cases), and tibia (n = 6 cases). Finally, those model-based transfer equations were compared to test-based transfer equations that were developed, using the same regression analyses, from limited published matched-condition tests of the same surrogates tested under nominally identical loading conditions.

METHOD

Validation of the FHBM

The FHBM began as an FE model of the human head (Ruan et al. 1993 1994). Over time, it evolved into a whole body human model which was extensively validated against published cadaveric test data from numerous blunt impact and whole-body frontal and lateral sled tests [Ruan et al. \(2003a, 2003b, 2005a, 2005b, 2006, 2008a, and 2008b\)](#), [El-Jawahri et al. \(2008, 2009, 2010a, and 2011\)](#), [Rouhana, El-jawahri and Laituri \(2010\)](#), and [Laituri and El-jawahri \(12\)](#). [El-Jawahri et al.](#)

(2010b) determined that the existing FHBM represented a 55-year old male. To investigate and account for the effect of age, two additional models representing 35- and 75-year old (YO) males were developed and validated. El-Jawahri, Laituri, and Kim (2012) further validated the three age-dependent models against several lateral and oblique thoracic and abdominal impact tests.

In the current study, the 55 YO version of the human model was further validated against knee blunt impact test data (Donnelly and Roberts, 1987). In those tests, axial impacts were performed on the right and left femurs of nine cadavers (62-year old $\pm$ 5.8). Load cells were implanted in the right femur of the test subjects to measure the internal force. The external force was calculated from the mass and acceleration of the impactor. In the current study, the tests were divided into four groups based on the impact speeds: 3.8, 6.5, 8.3, and 11.7 m/s. The mid-shaft section femur force and the external impact force were output from the model. The internal forces, external forces, and ratios of the internal force to the external force from the tests and the model were compared.

Validation of the HIII50 FE Model

Altair's Radioss Version 10 of the HIII50 FE deformable model was used to conduct the matched simulations. In the current study, the HIII50 model was validated against published tests of different loading conditions and severities. Those tests included: thoracic pendulum impacts (Yaguchi, Ono and Masuda 2008), rigid knee impacts (Donnelly and Roberts 1987, Masson 2002), padded knee impacts (Rupp et al. 2008), and frontal sled impacts (Forman et al. 2006).

Yaguchi et al. (2008) performed component level tests with the HIII50 dummy. The thoracic pendulum impacts at 4.3 and 6.7 m/s were chosen to validate the dummy model. The impact location was at the mid-sternum. The impactor was 23.4 kg and had a 152 mm diameter. The HIII50 model was used to simulate the tests. The force-deflection responses from the model were compared to the respective test data for both impact speeds.

Donnelly and Roberts (1987) subjected the HIII50 to knee blunt impacts at different speeds. The impactor was an 18 kg, 75 mm diameter rigid disc. The femur internal and external forces were measured. Similar to the cadaveric data, the HIII50 tests were divided into three groups subject to the impact speed: 1.51, 4.44, and 7.59 m/s. The HIII50 FE model was used to simulate those tests at the aforementioned speeds. The internal forces, external forces, and ratios of the internal force to the external force from the tests and the models were compared.

Masson (2002) used a 12 kg, 38 mm diameter horizontal rigid cylinder to perform blunt knee impacts to the HIII50. The tests were performed at 2.8 and 4 m/s speeds. The tests were

simulated with the dummy model. The femur axial force-time responses from the model were compared to the respective test data.

Rupp et al. (2008) performed padded symmetrical impacts to the knees of the HIII50. The total mass of the impacting structure was 255 kg. Three speeds were used: 1.2, 3.5, and 4.9 m/s. The padding material was Sorbothane. For the 1.2 m/s impacts, a 38 mm layer of 50-Durometer padding was used. For the higher speeds, a 25 mm layer of the 50-durometer padding was placed over a 25 mm layer of stiffer 70-durometer padding. The HIII50 model was used to simulate those tests. The femur axial force-time responses from the model were compared to the respective test data.

Forman et al. (2006) performed frontal sled impacts with the HIII50. The dummy was seated in a passenger seat of a midsized sedan restrained with a standard (not load-limited) 3-point belt. Two pulses with ΔV s of 29 and 38 km/hr were used. The HIII50 FE model was used to simulate the lower speed test. The head acceleration, chest deflection, and belt forces from the model were compared with the respective test data.

Matched-Condition Simulations

The validated HIII50 and the age-dependent FE models were subjected to matched-condition simulations in blunt impacts to the head, back, chest, knee, and foot. The simulated setups (impactor geometry and mass, impact speed, etc.) were similar to documented PMHS tests. The matched-condition impacts that were simulated with the HIII50 and the age-dependent FHBM FE models are discussed next.

Head Impact

Two types of head impacts were simulated: blunt impacts and drops onto a rigid surface. The whole model was used in the blunt impact cases. Each surrogate was oriented in an upright sitting posture such that the Frankfurt plane was horizontal. The impacts were 18 mm above the head center of gravity (CG) at the mid-sagittal plane for each surrogate. Rigid and padded 3.2 kg, 28 mm diameter, impactors were used. Eight and 16 mm thick Sorbothane foam was used in the padded cases. The impacts were simulated at velocities of 1, 2, 3, 4, 5, and 6 m/s. Figure 1 shows the head impact setup.

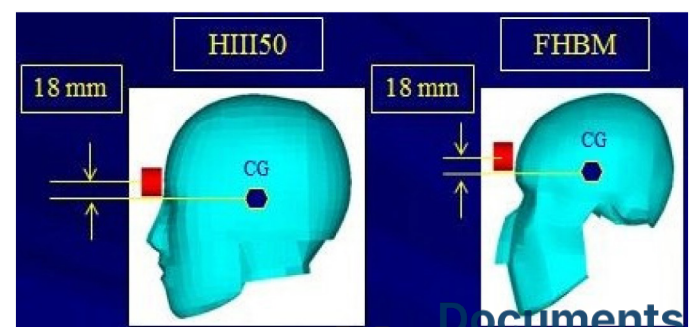


Figure 1. Head frontal Impact setup; HIII50 left, FHBM right.

Only the head was used in the drop simulations. The head was rotated so that the Frankfurt plane was at 15 degrees to the vertical. The head was dropped and initial contact occurred 30 mm above the head CG in the mid-sagittal plane. The impact speeds simulated were 0.6, 0.9, 1.2, 1.5, 1.8, 2.1, 2.4, 2.7, 3, and 3.5 m/s. [Figure 2](#) shows the head drop setup.

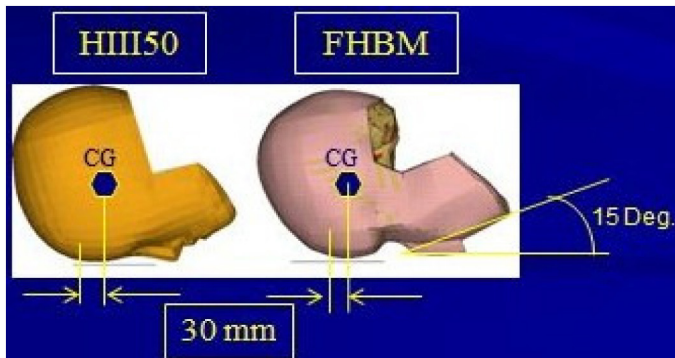


Figure 2. Head drop setup; HIII50 left, FHBM right.

A small sample of matched-condition tests with the HIII50 and cadavers were reported in the literature ($n = 6$). The tests involved frontal sled impacts ([Veizin et al. 2002](#), [Petitjean et al. 2002](#), and [Forman et al. 2006](#)). The surrogates were restrained with different restraint systems. More details can be found in the aforementioned references. The dataset was used to derive a test-based transfer equation using the same protocol used in the model-based transfer equation. The test- and model-based transfer equations were compared.

Back Impact

To generate neck responses, a 152 mm diameter, 23.4 kg rigid impactor was used to strike the back of the surrogates. The impact location was 260 mm below the head CG at the mid-sagittal plane. The impact speeds were 1, 2, 3, 4, 5, and 6 m/s. The upper neck load cell loads were the HIII50 model output. Those were corrected to calculate the loads at the Occipital-Condyle joint (OC). A section was created at the OC of the FHBM. The loads from that section were the PHMS model output. [Figure 3](#) shows the back impact setup.

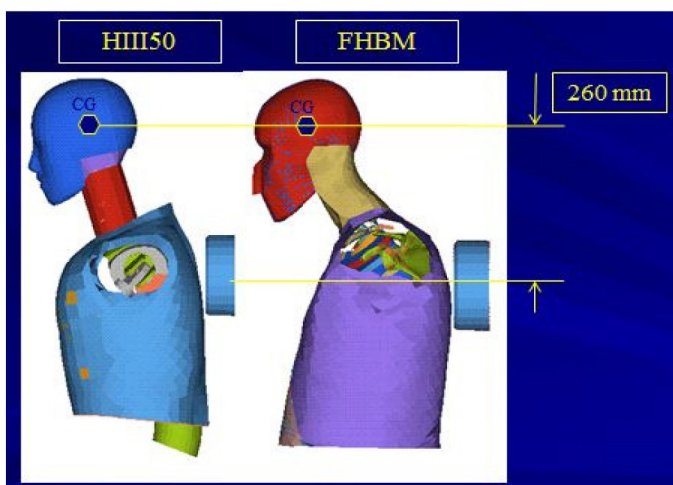


Figure 3. Back impact setup; HIII50 left, FHBM right.

Chest Impact

A 152 mm diameter, 23.4 kg rigid impactor was used in the thoracic impact simulations. The impact location was at the level of the fourth intercostal space at the mid-sagittal plane. The thoracic impacts were simulated at 1, 2, 3, 4, 5, and 6 m/s for each surrogate. The HIII50 chest deflection was the model output. The chest deflection of the HIII50 and age-dependent FHBMs were normalized to the chest depth at the impact location. The HIII50 chest depth is 229 mm ([Neathery et al. 1974](#)). The FHBM chest depth is 213 mm. [Figure 4](#) shows the thoracic frontal impact setup.

As mentioned earlier, [Laituri et al. \(2005\)](#) gathered a test-based dataset of matched-condition tests with the HIII50 and cadavers to develop their thoracic risk curve ($n = 20$). Those tests were performed under different loading conditions. More details on the test conditions could be found in [Laituri et al. \(2005\)](#). The same dataset was utilized in the current study to derive a test-based transfer equation using the same procedure used in developing the model-based transfer equation. The test- and model-based transfer equations were compared.

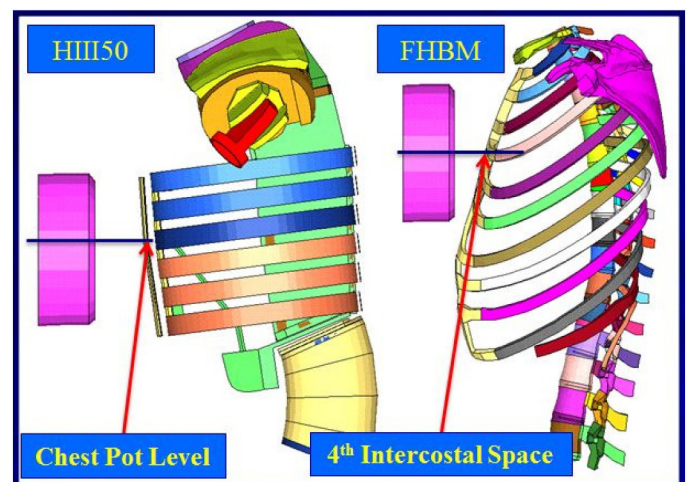


Figure 4. Thoracic frontal impact setup; HIII50 left, FHBM right.

Knee Impact

An 18 kg, 76 mm diameter impactor was used in the knee impact simulations. The femur was oriented horizontally for each surrogate. The leg was flexed downwards at 90 degrees. Rigid and padded impacts were used. Two padding thicknesses were used; 25 and 50 mm of the same Sorbothane foam that was used for the head impact simulations. The impact velocities simulated were 1, 2, 3, 4, 5, and 6 m/s. The femur force (F_z) of the HIII50 model and the axial force of two sections at the mid-shaft and distal end of the femur of the FHBM model were the output from the simulations. [Figure 5](#) shows the knee impact setup whereas; [Figure 6](#) shows the femur orientation of both surrogates.

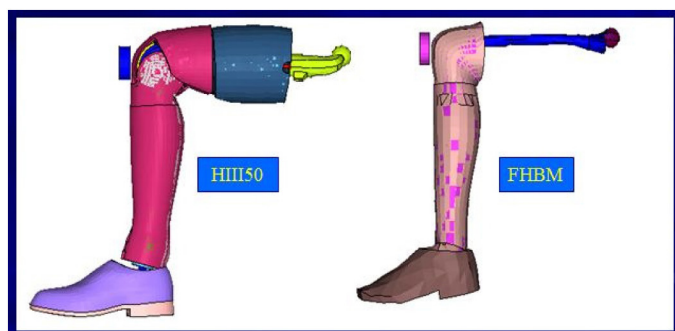


Figure 5. Knee impact setup: HIII50 left, FHBM right.

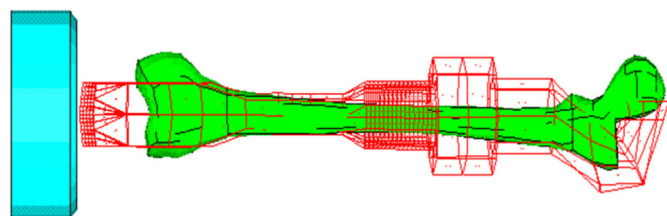


Figure 6. Femur orientation in knee impact: HIII50 red, FHBM green.

Foot Impact

In the foot impact simulations, each surrogate was seated on a rigid seat with the leg flexed at 30 degrees from vertical. The tibia orientation of the two surrogates was similar. A rigid planar surface of 9 kg mass representing a foot rest was used to impact the left foot along the long axis of the tibia at 1, 2, 3, 4, 5, and 6 m/s. The F_z and M_y from the upper and lower tibia load cells were the output from the HIII50 model. The loads from two tibia sections, at similar locations as the tibia load cells in the HIII50, were the output of the age-dependent human body models. Figure 7 shows the foot impact setup.

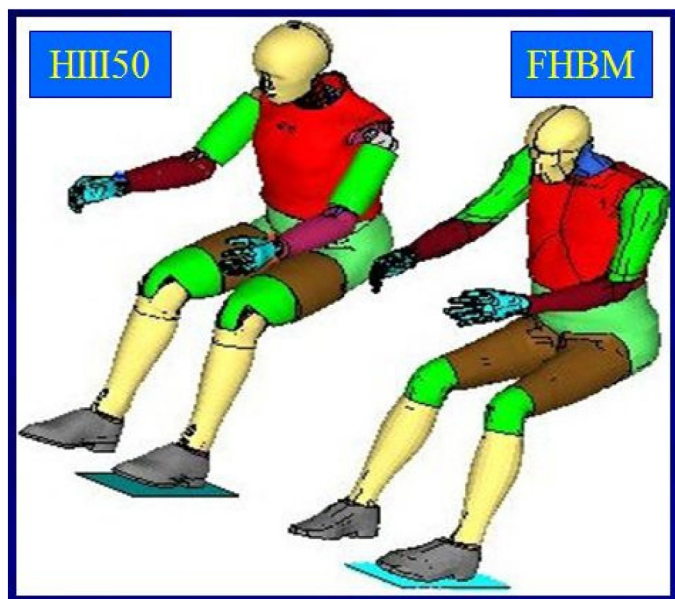


Figure 7. Foot impact setup; HIII50 left, FHBM right.

RESULTS

Validation of the FHBM

As mentioned earlier, the FHBM was extensively validated previously on both component and full system levels. In the current study, the 55 YO version of the FHBM was further validated against knee blunt impact tests (Donnelly and Roberts 1987). The tests were divided into four groups subject to the impact speed. Those cases were simulated. Table 1 summarizes the cadaveric and model responses.

Table 1. Comparison of Test versus CAE Responses, Cadaveric Knee Impact (Donnelly and Roberts 1987).

Speed m/s	Cadaveric Tests			FHBM		
	Femur	External	Ratio	Section	External	Ratio
	Force (kN)	Force (kN)		Force (kN)	Force (kN)	
3.80	4.87	9.76	0.50	3.67	7.61	0.48
6.54	8.96	14.63	0.59	7.13	13.41	0.53
8.30	9.34	14.39	0.65	8.53	15.33	0.56
11.70	12.03	22.11	0.54	11.71	20.49	0.57

Validation of the HIII50 FE Model

In the current study, Altair's HIII50 full FE Radioss model was validated against several sets of test data. They included thoracic blunt impact, knee rigid and padded impact, and frontal sled impact. The HIII50 model was also validated against the Donnelly and Roberts tests (1987). In their tests, the femur load cell force, external force, and ratio of internal to external forces were measured. Table 2 summarizes the femur load cell forces, external forces, and ratios of internal to external forces from the tests and simulations. Figures A1, A2, A3, A4, A5 show the comparison of the HIII50 model responses and the respective test data.

Table 2. Comparison of Test versus CAE Responses, HIII50 Knee Impact (Donnelly and Roberts 1987).

Speed m/s	HIII50 Tests			HIII50 Model		
	Femur	External	Ratio	Femur	External	Ratio
	Force (kN)	Force (kN)		Force (kN)	Force (kN)	
1.51	3.49	5.40	0.65	3.22	5.03	0.64
4.44	16.94	26.40	0.64	14.87	23.52	0.63
7.59	30.04	43.70	0.69	22.99	37.69	0.61